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Optical Correlation Tomography: Comparison of Imaging Resolution With and Without Ultrasound

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Abstract

The non-invasive recovery of various properties of biological tissues producing cross sectional images, named as tomography, has been used in detection and diagnosis of various diseases. In this paper, we demonstrate the improvement in resolution of the optical tomography by the introduction of focussed ultrasound, thereby localising the region of imaging. Towards this, we recover the amplitude of vibration of scatterers in the presence of ultrasound and the light absorption in the absence of ultrasound and compare the images with respect to the recovery resolution. Correlation Diffusion Equation, describing the diffusive propagation of autocorrelation of light in turbid media is used as the forward model for both of the cases.

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Keywords: Ultrasound modulated optical tomography;Correlation Diffusion Equation;Amplitude of vibration of scatterers.

1. Introduction

Medical imaging is a promising tool for the detection and diagnosis of various diseases. The earlier approach for imaging tissues was x-ray computed tomography which makes use of x-ray source. But owing to the harmful effects of x-ray, imaging methods that used other probing energy sources were explored wherever suitable. In optical tomography, Near Infrared radiation is used to image the tissue. Even though optical imaging methodologies received wide acceptance due to non-ionizing and non-harmful nature of light, there was an increasing demand for an imaging scheme that can provide a better spatial resolution. This was because of the highly scattering nature of light in biological tissues which limited its spatial resolution to optical mean free path.

Ultrasound Modulated Optical Tomography [1–3] combines the advantages of both optical source and ultrasound. When compared to light, ultrasound is less scattered by biological tissue and hence can be focused to a smaller region thereby increasing the spatial resolution. In UMOT, ultrasound is applied to the scattering medium to modulate the

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light inside the medium. The modulation of light by ultrasound was analytically modelled in [4–6]. The effect of ultrasound appears as a modulation in the intensity of the speckle pattern observed at the boundary.

The objective of this paper is to illustrate the resolution advantage of using ultrasound along with light, as done in UMOT, for recovering tissue properties from boundary measurements. In the presence of ultrasound we recover the amplitude of vibration of scatterers in the ultrasound focal volume from boundary measurements of light intensity autocorrelation [7]. The amplitude of vibration of scatterers depends on the stiffness of the material. Hence a measure of elasticity can be obtained by recovering the amplitude of vibration of scatterers. In the absence of ultrasound the optical absorption coefficient is recovered from the same measurements. For both of the cases i.e. with and without ultrasound, the recovery is performed using a forward model, which describes the diffusive propagation of amplitude autocorrelation of light in a turbid media, the Correlation Diffusion Equation [8,9]. The results illustrates the resolution advantage of ultrasound over light. The recovery of the tissue properties is performed by formulating it as a nonlinear least square minimization problem.

The paper is organized as follows. Section 2.1 provides a theoretical description of UMOT. The CDE describing the diffusive transport of amplitude autocorrelation through a tissue like media is given in section 2.2. Section 2.3 describes forward models for the recovery of tissue properties. To obtain forward models, a perturbation equation is formed from the CDE by considering ultrasound as a source of perturbation. Section 2.4 describes the iterative reconstruction algorithm, used for recovering the amplitude of vibration and the optical absorption coefficient. Simulation results are presented in section 3 and final concluding remarks are given in section 4.

Nomenclature

CDE	Correlation Diffusion Equation
CTE	Correlation Transport Equation
ROI	Region Of Interest
UMOT	Ultrasound Modulated Optical Tomography

2. Theoretical Background

2.1. Ultrasound Modulated Optical Tomography

Ultrasound Modulated Optical Tomography is a hybrid imaging method which combines the high contrast of optical imaging and higher resolution for ultrasound. In UMOT, ultrasound is focused into the object to be imaged to modulate the light inside the object. Application of ultrasound results in periodic compressions and rarefactions within the tissue, which introduces two optical effects: change in local refractive index and vibration of the scattering particles. Due to these effects, the photons passing through or near the ultrasound focal volume accumulate phase modulation and are called ultrasound tagged photons. These tagged photons carry information regarding the tissue at the region where light and ultrasound interact, named as the region of interest, and hence can be used to yield information about the tissue properties within the ROI. The ultrasound tagged photons on interference with untagged photons produce a speckle pattern at the boundary of the tissue. The effect of ultrasound appears as a modulation in the intensity of the speckle pattern, which is measured as a modulation in the light intensity autocorrelation. This measurement can be used to determine the change in refractive index and the amplitude of oscillation of scattering particles. Here we assume that the tissue is incompressible and hence the change in refractive index, Δn is negligible. Thus we make use of the boundary measurement of modulation depth in the intensity auto correlation of light, to recover the amplitude of vibration of scatterers within the ROI. The dependance of modulation depth on mechanical and optical parameters were investigated in [10]. It is found that the variation of modulation depth with mechanical parameters is sharper than its variation with local optical absorption μ_a.

In order to relate the measurements to the amplitude of vibration, one need a photon transport model which can account for the tissue dynamics. The diffusion of temporal field autocorrelation through a heterogeneous turbid media was studied in [11,12] and is found that position depended measurements of the field correlation can be used to image spatially varying dynamic properties. The transport of field correlation was modeled by a Correlation Transport

Equation. Later based on this model Wang et al [8,9] developed temporal Correlation Transfer Equation for multiply scattered light in Ultrasound modulated optical tomography. Here we use a diffusive approximation of the CTE i.e. Correlation Diffusion Equation, where the propagation of autocorrelation through the tissue is assumed to be diffusive.

2.2. The Correlation Diffusion Equation and the Forward Models

The CDE describes the diffusive propagation of light through the turbid media. The amplitude autocorrelation, $\phi(r,\tau)$ is the angle averaged form of the mutual coherence function of the field quantity, $I(r,k_s,\tau)$, which is dependent on the scattering angle represented by the propagation vector k_s. Basically the CDE is an elliptic partial differential equation with the autocorrelation of light $\phi(r,\tau)$ as the dependent quantity. In CDE it is assumed that $k_a l_{tr} \gg 1$ where k_a is the modulus of acoustic wave vector and l_{tr} is the transport mean free path of light. The CDE is given by

$$-\nabla.D\nabla\phi(r,\tau) + (\mu_a + \mu_s\hat{\varphi}(r,\tau))\,\phi(r,\tau) = S_0(r_0) \tag{1}$$

where

$$\hat{\varphi}(\tau) = \frac{1}{2}|P_a|^2 sin^2\left(\frac{\omega_a\tau}{2}\right)\left[\eta^2(k_a l_t r)tan^{-1}(k_a l_{tr}) + \frac{S_a^2}{3} - 2\eta S_a\right] \tag{2}$$

μ_a and μ_s are the absorption and scattering coefficients respectively and $S_0(r_0)$ is the isotropic source at $r_0 \in \Omega$. $D = 1/3(\mu_a + \mu_s)$ is the optical diffusion coefficient. The forward model comes with the boundary condition,

$$\phi(r,\tau) + D\frac{\partial\phi(r,\tau)}{\partial n} = 0, r \in \partial\Omega \tag{3}$$

where $\partial\Omega$ is the boundary of the domain Ω.

2.3. Recovery Of Amplitude Of Vibration and Optical Absorption Coefficient

2.3.1. Recovery of amplitude of vibration

In order to recover the amplitude of vibration of the scatterers, $p = |P_a|^2$ a nonlinear perturbation equation is developed which connects the ultrasound induced effects, which in our case is the amplitude of vibration of scatterers, to the change in intensity autocorrelation [7]. In the absence of ultrasound equation (1) becomes

$$-\nabla.D\nabla\phi(r,\tau) + (\mu_a + B(r,\tau))\,\phi(r,\tau) = S_0(r_0) \tag{4}$$

where $B(r,\tau)$ is the Brownian motion term, which is not taken in to account here. The boundary condition is given by

$$\phi + D\frac{\partial\phi}{\partial n} = 0 \tag{5}$$

Now we consider ultrasound as a source of perturbation, i.e. application of ultrasound results in a perturbation ϕ^δ in the intensity autocorrelation hence ϕ and becomes $\phi + \phi^\delta$. Hence equation (4) becomes

$$-\nabla.D\nabla(\phi + \phi^\delta)(r,\tau) + (\mu_a + B(r,\tau) + A(\tau)\chi_I p(r,\tau))(\phi + \phi^\delta)(r,\tau) = S_0(r_0) \tag{6}$$

Boundary condition is given as

$$(\phi + \phi^\delta)(r,\tau) + D\frac{\partial(\phi + \phi^\delta)(r,\tau)}{\partial n} = 0, r \in \partial\Omega \tag{7}$$

where

$$A(\tau) = \frac{1}{2}\sin^2(\frac{\omega_a\tau}{2})[\eta^2 k_a l_{tr}\tan^{-1}(k_a l_{tr}) + \frac{S_a^2}{3} - 2\eta S_a] \tag{8}$$

Here χ_I is the characteristic function of the insonified ROI, whose value is zero everywhere except for the ROI where it is unity. Combining both of the equations and neglecting the Brownian motion, we get a nonlinear perturbation equation connecting ϕ^δ and p, which is an elliptic partial differential equation in ϕ^δ as follows.

$$-\nabla.D\nabla\phi^\delta(r,\tau) + (\mu_a + A(\tau)\chi_I p)\,\phi^\delta(r,\tau) = A(\tau)\chi_I p\phi \tag{9}$$

with the boundary condition

$$\phi^{\delta}(r,\tau) + D\frac{\partial\phi^{\delta}(r,\tau)}{\partial n} = 0, r \in \partial\Omega \tag{10}$$

From the measured intensity autocorrelation at the boundary, we can deduce the amplitude autocorrelation and hence the ϕ^{δ} through the Siegert relation [13]. So the measurement quantity used for reconstructing p, the amplitude of vibration is the fourier transform of ϕ^{δ} evaluated at the ultrasound frequency $\omega = \omega_a$.

$$M(p, r, \omega_a)|_{r\epsilon\partial\Omega} = \int_0^{\infty} \phi^{\delta}(r,\tau)e^{-j\omega_a\tau}d\tau \tag{11}$$

Hence the UMOT problem of recovering p can be stated as, recovering the amplitude of vibration p in the insonified region I from the boundary measurement $\{M\}$. This is solved by formulating it as a least square minimization problem using an iterative reconstruction algorithm.

2.3.2. Recovery of Optical Absorption Coefficient

In order to show the resolution advantage of ultrasound over light, the optical absorption coefficient, μ_a of the tissue is also recovered in the absence of ultrasound using the same forward model. The CDE in the absence of ultrasound is described by

$$-\nabla.D\nabla\phi(r) + (\mu_a + B(r))\,\phi(r) = S_0(r_0) \tag{12}$$

with the boundary condition

$$\phi(r) + D\frac{\partial\phi(r)}{\partial n} = 0, r \in \partial\Omega \tag{13}$$

Here the unknown to be recovered is the optical absorption coefficient μ_a and the measurement available for reconstruction is the intensity autocorrelation $\phi(r)$ evaluated at the boundary. The μ_a is also recovered similar to the recovery of p, by formulating as a nonlinear least square minimization problem.

2.4. Iterative Reconstruction Algorithm

We need to determine the optical and mechanical properties of the tissue from a set of experimental measurements calculated at the boundary of the domain. The recovery of unknown tissue properties is performed using an iterative reconstruction algorithm. Fig. 1 shows the flowchart of the iterative reconstruction algorithm. The algorithm starts with an initial guess of the required property and is then refined iteratively. Usually this initial value is chosen as the background value of the normal tissue. Using this initial value, measurements are computed by solving the forward model. The value of the unknown is then updated by minimizing the mean squared error between experimental and computed measurements using Gauss Newton minimization. The process is repeated until the the error falls below some tolerable level.

For example in the case of the recovery of amplitude of vibration, let $\mathcal{M}$ be the operator which computes the measurements for a given value of p. Given the measurements $\mathcal{M}$, we need to recover the amplitude of vibration p. This can be stated as a minimization problem as follows:

$$\min_p \Theta(p) = \frac{1}{2}||\mathcal{M}(p) - M^e||^2 + \frac{\beta}{2}||p||^2 \tag{14}$$

Here the second term is the regularization term and $\beta > 0$ is the regularization parameter. M^e is the experimental measurements taken at the boundary $\partial\Omega$. Using Gauss-Newton method the update of p is calculated as

$$\Delta p = (J^T J + \lambda I)^{-1} J^T \Delta M \tag{15}$$

where $\Delta p = p^{(i+1)} - p^{(i)}$ and $\Delta M = M^c - M^e$ where M^c is the computed measurement and M^e is the experimental measurement. J is the Jacobian matrix which is the rate of change of measurement with respect to the parameter to be recovered. Here we compute Jacobian matrix in a computationally efficient way using the Adjoint and Green's function as described in [7].

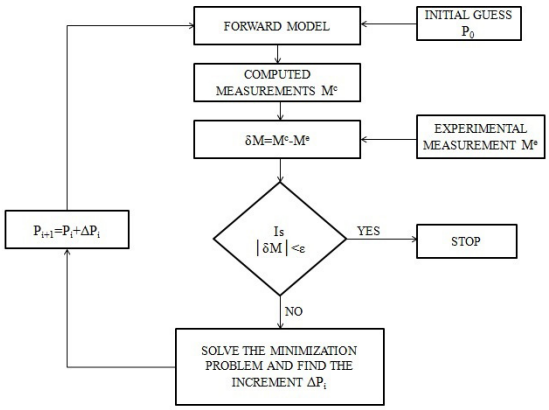
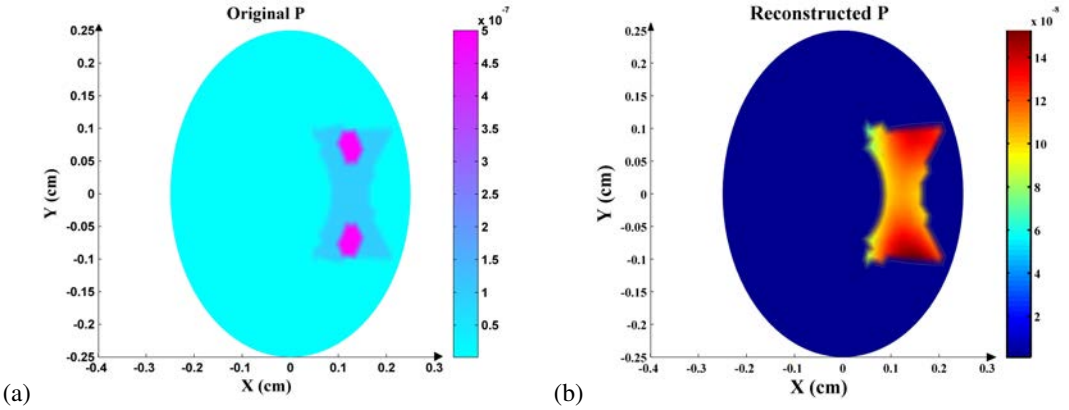


Fig. 1. Flowchart of the iterative reconstruction algorithm.

3. Simulation Results

3.1. Recovery Of Amplitude Of Vibration

Here experimental measurements are numerically simulated with added noise . The geometry of the object considered for the simulation is 2-D and is a circular disc of radius 0.25 cm. A two dimensional hyperboloid of height 0.2 *cm* centered at (0.125 *cm*, 0 *cm*) is considered as the ultrasound insonified region. The background optical properties of the tissue is selected as μ_a 0.001 cm^{-1} and μ_s 20 cm^{-1}. The value of amplitude of vibration in the insonified region is chosen as $1 \times 10^{-7} cm^2$ and $1 \times 10^{-9} cm^2$ elsewhere. We assumed two circular inhomogeneities of radius 0.025 *cm* centered at (0.125 *cm*, 0.07 *cm*) and (0.125 *cm*, -0.07 *cm*) respectively. The value of amplitude of vibration in the inhomogeneous region was set as $5 \times 10^{-7} cm^2$.



(a) (b)

Fig. 2. (a) Assumed distribution of $p(r)$; (b) Reconstructed $P(r)$.

To simulate experimental measurements, the perturbation equation (9) is solved with this distribution of p for different values of τ varying from 0 to 5×10^{-7} with a step size of 5×10^{-8}. The resulting $\phi(r, \tau)$ is time fourier transformed and the component at the ultrasound frequency $\omega_a = 1MHz$ is determined. To mimic the experimental measurement 1% Gaussian noise is added to these measurements. Now we repeat this procedure for all the source positions by rotating the source detector arrangement. Here 18 different source positions are used with 17 detectors corresponding to each source position. Thus our experimental measurement consist of 18 set of 17 measurements. The Finite Element Method is used to solve the perturbation equation (9). For simulating experimental measurements a finer mesh with 1243 nodes and 2376 triangular elements is used. The iterative reconstruction algorithm described

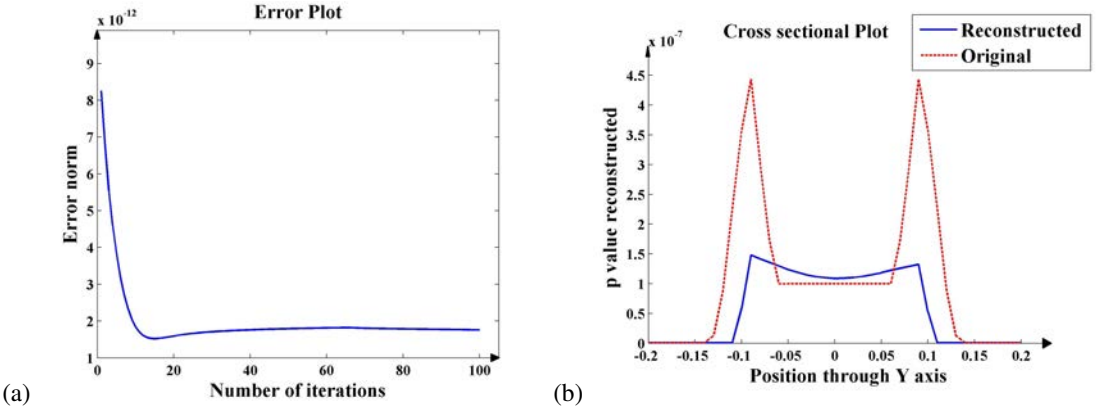


Fig. 3. (a) Measurement Error in various iterations; (b) Cross sectional plot through the center of inhomogeneity.

in section 2.4 is used for recovering p. The initial value of p is chosen as $1 \times 10^{-7} cm^2$, the background value for normal tissue. Using this, measurements are computed similar to experimental measurements except that a coarser mesh with 901 nodes and 1710 triangular elements is used. The update for p is calculated using equation (15) and is updated for each source position. This is repeated for several iterations until the error falls below a tolerable level. The value of regularization parameter β is determined by trial and error. Fig. 2 shows the original distribution of p used for simulating experimental measurements and the reconstruction obtained. The algorithm is able to reconstruct around 30% of the actual value. The values of p along the center of the inhomogeneities is plotted in Fig. 3(b). The corresponding values for original distribution is also given for comparison. A plot showing the norm of error between computed and experimental measurements in different iterations is shown in Fig. 3(b).

3.2. Recovery Of Optical Absorption Coefficient

The optical absorption coefficient of the tissue μ_a is recovered in the absence of ultrasound. Here simulation is done for two different cases. In case I the object used for simulation is same as that used for the recovery of amplitude of vibration, a circular disc of radius 0.25 cm with the background optical scattering coefficient μ_s 20 cm^{-1}. Two circular inhomogeneities of radius 0.025cm are assumed at positions (0.125 cm, 0.07 cm) and (0.125 cm, -0.07 cm) respectively. The background value of μ_a is chosen as 0.001 cm^{-1} and 1 cm^{-1} for the inhomogeneous regions. To simulate experimental measurements the forward model in equation (12) is solved with this distribution of μ_a and 1% Gaussian noise is added to these measurements. The effect of Brownian motion is not accounted in any of the simulations.

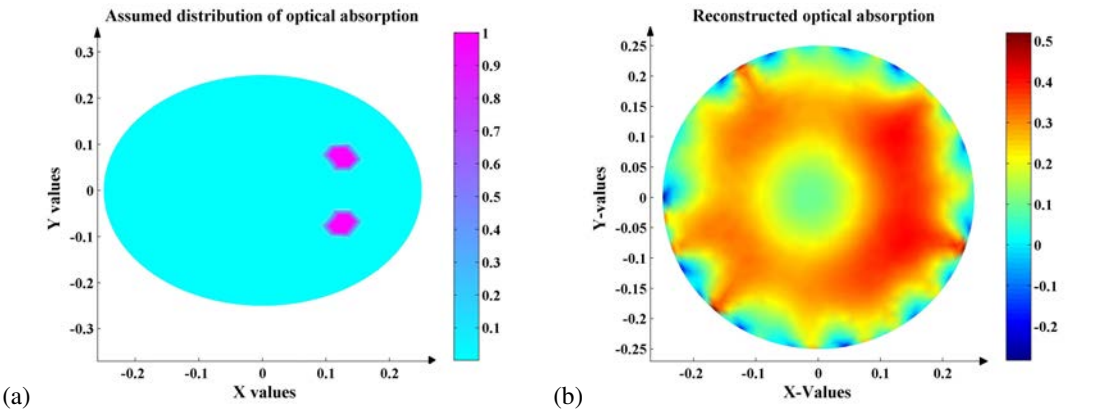


Fig. 4. (a) CASE I Assumed distribution of μ_a; (b) Reconstructed μ_a.

The reconstruction procedure is same as that for the amplitude of vibration. We start with an initial value of μ_a 0.001 cm^{-1}, the background value for the normal tissue. μ_a is then iteratively updated using Equation (15). Here the update is calculated for the entire tissue cross section, whereas previously with ultrasound, the update was calculated only for the hyperboloid region due to presence of χ_I, the characteristic function of the ROI. The mesh densities and the source-detector arrangements used are same as that in UMOT. In case II the object used for simulation is a circular

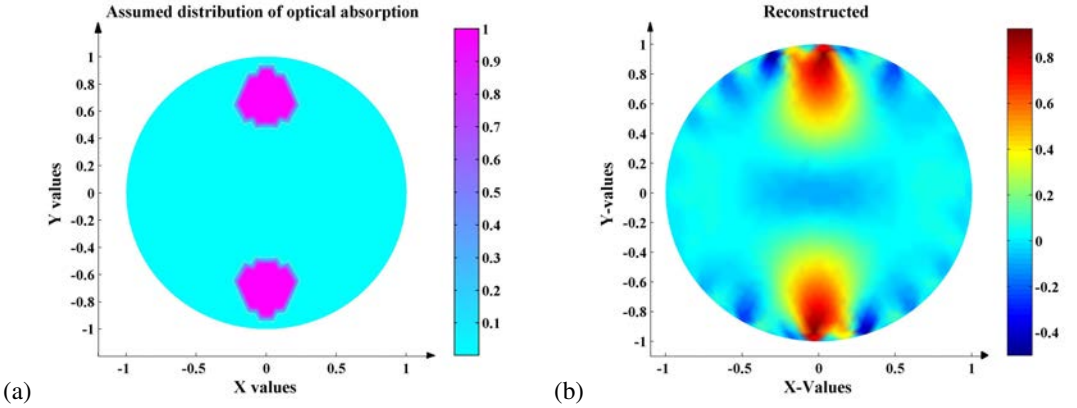


Fig. 5. (a) CASE II Assumed distribution of μ_a; (b) Reconstructed μ_a.

disc of radius 1 *cm* with the background optical scattering coefficient μ_s 20 cm^{-1}. Two circular inhomogeneities of radius 0.2 *cm* are assumed at positions (0 *cm*, 0.7 *cm*) and (0 *cm*, -0.7 *cm*) respectively. The background value of μ_a is chosen as 0.001 cm^{-1} and 1 cm^{-1} for the inhomogeneous regions. The simulation of experimental measurements and reconstruction procedure is same as previous. Fig. 4 shows the original distribution of μ_a used for simulating experimental measurements and the corresponding reconstruction obtained for case I. Fig. 4 is the original distribution and the reconstruction obtained for case II. Clearly for case I the algorithm fails to resolve between the two inhomogeneities. But for the same simulation object, with ultrasound we were able to resolve and locate the two inhomogeneities with satisfactory accuracy. This reveals the resolution advantage of ultrasound over light. For case II, we used a larger simulation object and larger dimension for the inhomogeneity also the two inhomogeneities were placed far part. Clearly for case II the two inhomogeneities are well resolved and are located correctly. For this case the algorithm is able to recover more than 80% of the actual value. Thus it is possible to achieve good reconstruction with light source, but the resolution that can be achieved with it is limited. Thus one cannot resolve inhomogeneities of dimension less than the optical mean free path using light as the only source.

4. Conclusion

In this work we have presented a comparison of imaging resolution with and without ultrasound. The amplitude of vibration of scatterers, p in soft biological tissues is recovered in the presence of ultrasound from boundary measurements of light intensity autocorrelation. Even though the contrast as well as the accuracy was limited, the method was able to locate the position of the inhomogeneities in p within the tissue. In order to show the resolution advantage of ultrasound, the optical absorption coefficient of the tissue was also recovered without using ultrasound for two different cases. The method failed to resolve the two inhomogeneities when they were small and closely positioned, whereas good reconstruction was obtained for two larger inhomogeneities when placed far apart. This clearly proves that with light as the only source, the amount of spatial resolution that can be achieved is limited whereas with light and ultrasound one can achieve better spatial resolution. In the present work we have recovered the amplitude of vibration of scatterers as a measure of mechanical property, as it depends on the stiffness of the tissue and can provide an indication of the stiffness of the tissue.

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